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Thin-film micro-coil detectors: Application in MR-thermometry

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ABSTRACT

Thin-film micro-coils are integrated with commercially available ablation catheters, for MR-thermometry during laser interstitial thermal therapies (LITTs). The coils are formed on a flexible polyimide substrate and consist of a two-turn electroplated copper inductor and integrated parallel plate capacitors for tuning and matching. Their performance was assessed during Nd:YAG laser ablations in a static phantom study carried out in a 3 T clinical scanner. Further moving phantom studies were performed to calculate errors due to motion. The temperature accuracy is improved by 1.5–10 times in a radius matching the dimensions of the lesions typically treated. Resolution of 1 mm can be maintained during motion by using short acquisition time sequences while the SNR remains sufficient for accurate MR-thermometry. The temperature error on a moving un-heated phantom under respiratory gating does not exceed 1 °C. This demonstration suggests a possible improvement of the overall control of MR-guided LITTs by local temperature monitoring.

1. Introduction

Micro-coils have received considerable attention from the magnetic resonance imaging (MRI) and spectroscopy communities, for biosensing, tracking of interventional devices, intravascular and internal organ imaging and MR-guided thermal therapies [1–8]. The increase in the signal-to-noise ratio (SNR) resulting from close coupling to the signal source, compared to volume and surface coils [9,10], makes micro-coils useful for fast, high resolution imaging, when a small field-of-view (FOV) can be tolerated.

Lack of repeatability and the effort required for matching and tuning using discrete components have limited clinical use. In recent years, micro and nano-fabrication techniques and the use of flexible substrates have opened up new design avenues [4,5,11–16], allowing the batch fabrication of disposable coils with repeatable coil parameters. Additionally, intense research effort has been invested in the safety of micro-coil-based catheters and many designs for improved patient safety have been reported [17]. We have recently presented thin-film detectors with mechanically tunable integrated capacitors [18], with potential application in the early diagnosis and staging of biliary carcinoma [19].

Here we explore a different application for thin-film micro-coils: improvement in the control of MR-guided laser interstitial thermal therapies (LITTs). Such minimally invasive therapies are increasingly used to destroy lesions in the liver and efforts are being made to improve their efficacy [20]. A key aim is to improve temperature accuracy during MR-thermometry [21]. High SNR is critical to accurate, real-time temperature monitoring, and can allow precise calibration of the MR thermal coefficients and quantification and exclusion of non-temperature related effects. Moreover, it can ensure that the impact of motion artefacts is minimised, by allowing faster sequences that still give useful SNR.

To date, only a few systems combining ablation devices with internal coils [8,22,23] have been proposed. None offers a disposable solution and no system has been reported for use during MR-guided LITTs, even though this is a well-established procedure for the treatment of small (<20 mm diameter), inoperable lesions. The paper is organised as follows: Section 2 describes the design, fabrication, and electrical performance of thin-film micro-coil detectors and details their integration with commercially available laser applicators. Section 3 describes the experimental set-up and phantoms used to assess their performance. Section 4 presents a comparison between the performance of the micro-coil system and an external array coil. Finally, discussions and conclusions are presented in Section 5.

2. Thin-film micro-coil detectors

In this Section, we describe the integration of a micro-coil with an ablation catheter. We outline the resonator design, the

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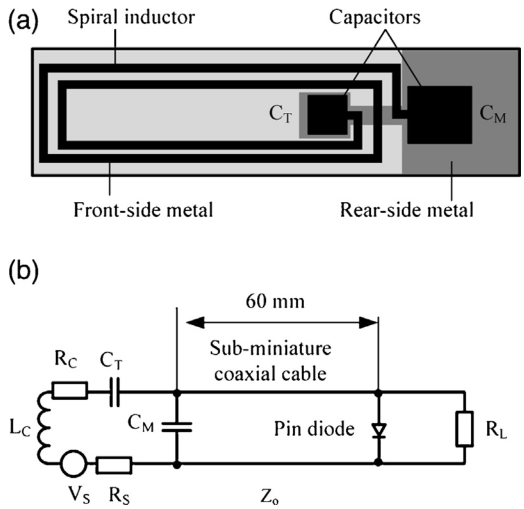

Fig. 1. (a) Plan view of the thin-film micro-coil detector, (b) electrical equivalent circuit.

fabrication procedure and describe a method for identifying component values for matching and tuning at 127.6 MHz. Electrical performance is assessed using a network analyser (E5061A, Agilent Technologies).

2.1. Design and principle of operation

The micro-coil receiver is shown in Fig. 1a. It consists of a copper-clad Kapton thin film, patterned to form a two-turn spiral inductor with a pair of integrated capacitors C_T and C_M for tuning and matching. The inductor is connected to the two capacitors, each of which uses the substrate as an interlayer dielectric. The front side pattern consists of a spiral linked to two plates, while the rear side pattern consists of a pair of plates linked directly together. This layout places C_M outside the coil, allowing direct connection to a coaxial cable.

The layout can be fabricated from patterned conductors since no air-bridge is needed to exit the spiral. Double-sided processing is required but front-to-back alignment is not critical since capacitor plates need only overlap. Coils were designed with the following parameters: conductor width 150 μm, conductor separation 100 μm, coil length 60 mm, and coil width 4.5 mm. The last value was chosen to place the long conductors approximately on the diameter of the ablation catheter.

Fig. 1b illustrates the electrical equivalent. A shunt matching circuit has been preferred to avoid vias. For maximum power transfer, the signal induced by the external nuclear dipoles should be matched to a real load R_L at angular frequency ω_S using the capacitors C_M and C_T. The complex impedance of the parallel combination of R_L and C_M is $(1/j\omega \cdot C_M)\{1/(1+1/j\omega \cdot C_M R_L)\}$. If $\omega \cdot C_M R_L \gg 1$, this result may be approximated as $R_L' = 1/j\omega \cdot C_M + 1/(\omega^2 \cdot {C_M}^2 R_L)$. Matching requires a C_M such that $R_L' = R_C$ at the signal frequency:

$$C_M = 1/\{\omega_S \sqrt{(R_C R_L)}\} \quad (1)$$

Tuning involves choosing C_T so that the circuit is resonant at ω_S, which requires that the effective capacitance C_{eff} satisfies Eq. (2) with C_T positive, which in turn requires $\omega_S L_C > \sqrt{(R_C R_L)}$. C_M and C_T can be found by iteration after R_C, L_C have been determined. Both R_C and L_C can vary considerably without a well-controlled fabrication process.

$$1/C_{eff} = 1/C_M + 1/C_T = \omega_S^2 L_C \quad (2)$$

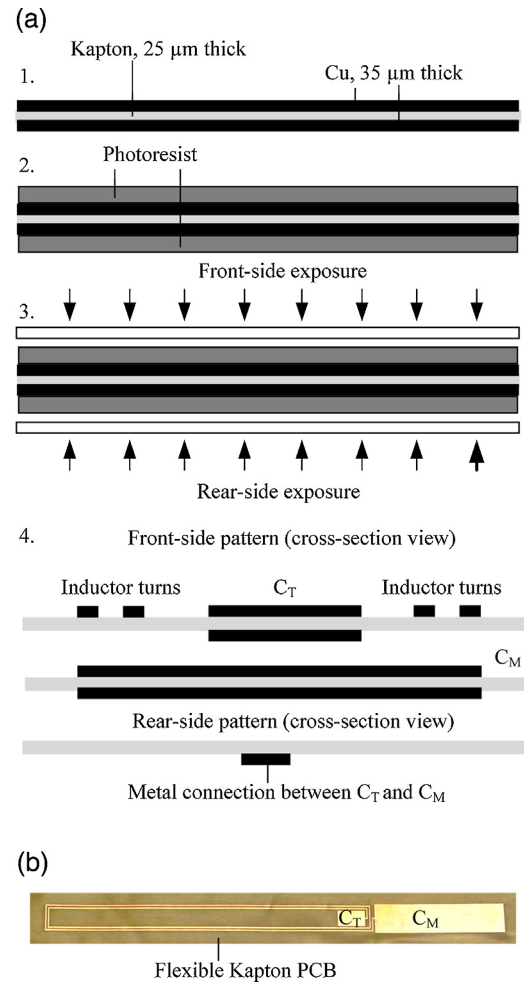

Fig. 2. (a) Process flow for detector fabrication, (b) batch-fabricated micro-coil with integrated tuning (C_T) and matching (C_M) capacitors.

2.2. Fabrication

Prototype devices were developed for 3.0 T ^{1}H MRI. Batch fabrication was carried out by the UK company Clarydon (Willenhall, West Midlands). The starting material is a 25 μm thick polyimide (Kapton®, HN, DuPont High Performance Films) carrying a 35 μm thick layer of Cu on either side. Patterning was carried out using double-sided exposure to a pair of photomasks formed from Mylar-coated silver halide on a 175 μm thick polyester backing. Masks were fabricated from a Gerber file.

The fabrication process is illustrated in Fig. 2a. Each side of the printed circuit board (PCB) (1) is coated with a 175 μm thick layer of laminated photoresist (2). The sensitised PCB is then sandwiched between the two photomasks on a glass backing using a set of pins passing through mating holes in each component for alignment. Textured Mylar spacers are used to assist evacuation. Each side of the PCB is exposed with an UV lamp (3). Resist development and metal etching are carried out with the PCB horizontal, using a leader board to allow dragging through a spray developer and etcher. The resist is then stripped (4). A fabricated device is shown in Fig. 2b.

L_C and R_C were determined in an iterative process to choose the correct areas for C_M and C_T prior to the design of the photomasks. A set of inductors was first fabricated without capacitors (Batch 1), but with bond-pads to allow addition of a wire-bond air-bridge and surface-mount capacitors, which were adjusted to match to 50 Ω and tune to 127.6 MHz. A second set of devices was then fabricated with integrated capacitors (Batch 2), using areas for